HIGH SPEED MODULATION OF SWITCHED-FOCUS X-RAY TUBE

The following relates to the diagnostic imaging arts. It finds particular application in conjunction with dose-modulation by cascading frequency responses of x-ray tube filament current and grid voltage for computed tomography
5   imaging, and will be described with particular reference thereto. However, the following relates more generally to dose-modulated and other types of computed tomography imaging and to x-ray tubes used in other applications.

In a typical computed tomography imaging apparatus,
10   an x-ray tube is mounted on a rotating gantry that defines an examination region inside which an imaging subject is disposed. The x-ray tube rotates about the subject on the rotating gantry and projects a wedge-, fan-, conical-, or otherwise-shaped x-ray beam through the examination region. A two-dimensional
15   x-ray detector disposed on the rotating gantry across the examination region from the x-ray tube receives x-ray beam after passing through the examination region. Suitable electronics estimate x-ray absorption data based on the detected x-ray intensities, and an image reconstruction
20   processor reconstructs an image representation based on the absorption data.

The x-ray tube in the above arrangement typically includes a filament that generates electrons. A cathode cup partially surrounds the filament and is biased negatively to
25   focus the electrons into an electron beam. A bias in the kilovolt range between the cup and a rotating anode accelerates the electron beam to the anode, causing it to emit x-rays.

A problem arises in that the imaging data is under-sampled in the rotational direction. Sampling occurs at a
30   spatial frequency related to the detector spacing, while sampling theory calls for a doubled spatial frequency to avoid aliasing and other sampling-related artifacts. To counteract under-sampling, an x-ray tube is employed in which the focal spot is alternated or wobbled between two discrete positions in

the rotational direction between measurements to spatially interleave samples. Alternatively, a quarter ray offset can be employed in which rays from opposing 180° projection views are spatially interleaved.

5   To effect beam wobble, the x-ray tube includes grid electrodes arranged on opposite sides of the filament. The grid electrodes are biased with an alternating polarity to generate a switched electrostatic force orthogonal to the electron beam that shifts the electron beam between two paths corresponding
10   to the two focal spots of the wobble. Alternatively, an orthogonal electromagnetic force can be used to switchingly steer the beam.

Another concern in computed tomography imaging is limiting x-ray exposure of the subject. In medical imaging
15   applications, the x-ray dosage delivered to the patient is a regulated safety parameter. In airport security scanning, the x-ray intensity is advantageously adjusted to account for differing x-ray absorption characteristics of different types of luggage. Various dose-modulated computed tomography
20   techniques have been developed. However, these past methods have certain disadvantages.

In one method, described in U.S. Patent No. 5,867,555 issued to Popescu et al., dose modulation as a function of angular position is obtained by synchronously modulating the
25   x-ray tube filament temperature. As noted in that reference, however, the modulation index obtained by filament current control is limited by a cooling rate (thermal mass) of the filament, especially at higher rotation speeds. The Popescu technique changes the dose once every 180° in a scanner that
30   rotates once every 0.75-2.0 seconds, which pushes the limit of the filament's ability to respond. Thermal mass limitations are also exaggerated for higher dynamic modulation ranges which involve substantial cooling of the filament. The cooling rate decreases as the temperature decreases.

35   Modern scanners rotate twice per second and faster speeds are starting to appear in product. Moreover, more dose changes per revolution are advantageous, setting modulation

speeds that exceed the physical limits of the Popescu technique.

In another known method, a Wehnelt cylinder incorporated into the x-ray tube selectively pinches off the beam to switch the beam on and off. Beam switching is synchronized with a CINE frame sequence to dose-modulate by controlling a duty cycle of the x-ray beam. However, this dose modulation approach reduces the amount of collected angular data.

The present invention contemplates an improved apparatus and method that overcomes the aforementioned limitations and others.

According to one aspect, a dose-modulated irradiating system for an x-ray tube is disclosed. The x-ray tube has a cathode including a filament that generates electrons which are focused into a beam, and an anode that generates x-rays responsive to the electron beam. At least one electrostatic control electrode is arranged to electrostatically reduce an intensity of the electron beam. A biasing means is provided for applying a time-varying electrical bias to the electrostatic control electrode to vary the intensity of the electron beam.

According to another aspect, a method is provided for dose-modulating an output of an x-ray tube. The x-ray tube includes a cathode having a filament that generates electrons which are focused into a beam, an anode that generates x-rays responsive to the electron beam, and an electrostatic control electrode that electrostatically adjusts an intensity of the electron beam. A time-varying electrical bias is applied to the electrostatic control electrode to produce a first time-varying intensity modulation of the electron beam.

One advantage resides in providing a high dynamic range of dose modulation at a high modulation frequency. This facilitates dose modulation using present- and next-generation computed tomography scanners which rotate at rates of 120 rpm or higher.

Another advantage resides in providing such dose modulation without modifying the tube design.

Yet another advantage resides in providing such dose modulation without additional high speed electronics.

Numerous additional advantages and benefits will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiments.

The invention may take form in various components and arrangements of components, and in various process operations and arrangements of process operations. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 shows a dose modulated computed tomography imaging apparatus.

FIGURE 2 schematically shows a suitable dose-modulated irradiating system employing electrostatic beam wobbling.

FIGURE 3 shows schematically shows a suitable dose-modulated irradiating system employing electromagnetic deflection beam wobbling.

With reference to FIGURE 1, a computed tomography imaging scanner 10 includes a subject support 12 for moving a subject such as a medical patient, an item of luggage undergoing a security scan, or the like into or within an examination region 14 defined by a rotating gantry 16. An x-ray tube 20 arranged on the gantry 16 transmits a fan-shaped, wedge-shaped, conically-shaped, or otherwise-shaped x-ray beam into the examination region 14. A two-dimensional x-ray detector 22 disposed on the gantry 16 across the examination

region **14** from the x-ray tube **20** measures a spatially-varying intensity of the x-ray beam after the x-ray beam passes through the examination region **14**. Typically, the x-ray detector **22** is mounted on the rotating gantry **16**. In another suitable arrangement, the detector is arranged circumferentially on a stationary gantry surrounding the rotating gantry. Gantry rotation is controlled by a gantry rotation controller **24**.

In helical computed tomography imaging, the gantry **16** rotates simultaneously with a linear motion of the subject support **12** to effect a helical trajectory of the x-ray tube **20** about the examination region **14**. In axial computed tomography imaging, the gantry **16** rotates while the subject support **12** remains stationary to effect a circular trajectory of the x-ray tube **20** about the examination region **14**. In volumetric axial imaging, the subject support **12** is repeatedly stepped linearly with an axial scan performed for each step to acquire multiple image slices along the axial direction.

Acquired imaging projection data with an index of the apex of the fan or cone and of the trajectory within the fan or cone is transmitted off the gantry **16** and stored in a digital data memory **30**. A reconstruction processor **32** reconstructs the acquired projection data, using filtered backprojection or another reconstruction method, to generate a three-dimensional image representation of the subject or of a selected portion thereof, which is stored in an image memory **34**. The image representation is rendered or otherwise manipulated by a video processor **36** to produce a human-viewable image that is displayed on a graphical user interface **38** or another display device, printing device, or the like for viewing by an operator. Preferably, the graphical user interface **38** is programmed to interface a radiologist with the computed tomography scanner **10** to allow the radiologist to execute and control computed tomographic imaging sessions.

During imaging, it is advantageous to modulate a x-ray radiation dosage received by the subject. For non-cylindrical objects, it is advantageous to have a higher x-ray beam intensity along the major axis and a lesser intensity along the minor axis. For example, in rapid airport security scanning of luggage the x-ray beam intensity is preferably adjusted depending upon the x-ray absorption characteristics of the scanned luggage. In medical imaging applications, the intensity is preferably adjusted to maintain a constant radiation dose for each axial slice or each helical rotation as an imaging scan passes through regions of the body having differing x-ray absorption densities. Present computed tomography scanners provide rotation rates of about 120-150 rpm, and rotation rates of 180-240 rpm and higher are contemplated in the foreseeable future.

A 120 rpm rotation rate corresponds to an acquisition time of 0.25 seconds for a 180° angular span of data, while a 180 rpm rotation rate corresponds to an acquisition time of 0.167 seconds for 180° of data. Hence, the dose modulation should have a frequency response of about 4 Hz or higher for a 120 rpm gantry rotation rate, and should have a frequency response of about 6 Hz or higher for a 180 rpm gantry rotation rate, to provide dose modulation that tracks the gantry rotation. Higher rotation speeds produce proportionately higher anode milliampere modulation rates. For medical imaging applications, a dose modulation dynamic range of at least about 8:1 to 10:1 is desired. That is, a ratio of a maximum x-ray beam intensity to a minimum x-ray beam intensity during the modulation at all rotation speeds should preferably be at least about 8:1 to 10:1.

To effect dose modulation conforming to these specifications, a dose delivery rate processor 50 integrates the x-ray absorption data over an area of the x-ray detector 22 and inputs a continuously updated dose delivery rate to a dose modulation controller 52. The gantry rotation controller 24 also inputs gantry angular position to the dose modulation

controller **52** for optional synchronization of the dose modulation with angular position of the x-ray source **20**.

   With continuing reference to FIGURE 1, the dose modulation controller **52** outputs two synchronized, cooperating
5 dose modulation control signals: a filament modulation signal **60** directed toward an x-ray tube filament power supply **62**, and a grid modulation signal **64** directed toward a grid power supply **66**. The grid power supply **66** additionally receives a beam wobble input signal **70** which alternates or wobbles a focal spot
10 of the x-ray tube **20** to counteract under-sampling and aliasing in the radial direction. The x-ray tube **20** is driven and controlled by outputs of the filament power supply **62**, the grid power supply **66**, and a tube power supply **72** which sets the potential between the cathode and the anode.
15   With continuing reference to FIGURE 1 and with further reference to FIGURE 2, which schematically shows components of the x-ray tube **20** in cross-section along with associated electronic components, an elongated filament **80**, preferably having a wire diameter, for a conventional coiled
20 wire filament, of 8-10 mil, is arranged in a cathode cup **82** of a cathode **84**. The filament **80** generates electrons by thermionic emission, field emission, or another mechanism in response to a filament current input delivered by the filament power supply **62**. Preferably, an isolation transformer **86** and insulating
25 standoffs (not shown) electrically isolate the filament **80** from other elements of the x-ray tube **20**. Typically, the filament current is a pulse-width modulated current with a frequency of about 5 kHz to about 30 kHz and an amplitude ranging from a few hundred milliamperes to a few amperes.
30   With reference to an electrical common **90**, the cathode **84** is typically biased negatively at a bias of around -40 kV to -70 kV and the anode to +40 kV to +70 kV for a bipolarly biased tube. The cathode cup **82** is shaped such that

the negative charge defines an electrostatic electric field that collimates electrons generated by the filament **80**. The electrons of the electron beam accelerate across the cathode-to-anode·gap, which is typically about 2 cm.

5   In response to the accelerated electron beam **94** striking the anode **92**, the anode emits an x-ray beam **96** that passes out of a vacuum volume (not shown) containing the filament **80**, the cathode **84**, and the anode **92**. The anode **92** rotates about an axis **100** and includes an angled

10   circumferential x-ray generating surface **102** that interacts with the electron beam **94** as the anode **92** rotates. The rotation distributes heat across the circumferential surface **102**.

         With continuing reference to FIGURES 1 and 2, the cathode cup of the x-ray tube **20** further includes first and

15   second grid electrodes **110, 112** arranged on opposite sides of the filament **80** such that the electron beam **94** passes between the grid electrodes **110, 112**. The grid electrodes **110, 112** are used to electrostatically modify the electron beam **94**.

         Specifically, biasing both electrodes **110, 112** more

20   negative narrows a width of the electron beam **94** by electrostatic constriction or aperturing, which affects the focus at the anode. A relatively larger negative sum voltage applied to both electrodes **110, 112** further reduces an intensity of the electron beam **94** at the anode **92** by partial

25   electrostatic pinchoff of the electron beam **94**, i.e. reduces the tube current. For a sufficiently large negative bias voltage applied to both electrodes **110, 112** the electron beam **94** can be pinched off entirely.

         Additionally, an applied voltage difference, imposed

30   upon the voltage sum, and between the grid electrodes **110, 112**, effects a beam deflection.

         In another embodiment, the voltage to each grid may be produced as the sum of individual voltages for constriction

and deflection, such that the focal spot or spots may be jointly moved essentially tangentially to the rotating anode surface while maintaining correct spot to spot displacement and width.

5   Additional elements such as one or more additional filaments for redundancy or for achieving different electron beam characteristics and the like, (elements not shown) are also optionally included in the cathode assembly.

        With continuing reference to FIGURES 1 and 2, the
10 filament power supply **62** is controlled by the filament modulation signal **60** produced by the dose modulation controller **52**. Modulation of the filament current causes modulation of the filament temperature and corresponding modulation of the electron generation rate. This, in turn, modulates the
15 intensity of the electron beam **94** for dose modulation. Changing the filament current does not significantly modify electron trajectories and so does not greatly alter the focus of the electron beam or change the size of the focal spot. However, the filament **80** has a thermal mass which limits the frequency
20 response of this modulation control. For a preferred filament having a 200 micron (8 mil) diameter, modulating the filament current at 4 Hz achieves the frequency response of dose modulation of 4 Hz with corresponding dynamic range of about 2:1 to 3:1. A larger diameter filament has more thermal mass
25 and will typically exhibit a slower frequency response and reduced dynamic range.

        In order to obtain dose modulation with better frequency response than that achieved using filament control alone, the dose modulation controller **52** additionally outputs
30 the grid modulation signal **64** directed toward a grid power supply **66**. The grid modulation signal **64** is combined with the beam wobble input signal **70** which alternates or wobbles a focal

spot of the x-ray tube **20** to produce combined grid electrode biasing potentials.

       Specifically, a first summer **120** additively combines the grid modulation signal **64** and the beam wobble input signal **70** to produce a first control signal that controls the electrostatic potential that is applied to the first grid electrode **110**. A second summer **122** subtractively combines the grid modulation signal **64** and the beam wobble input signal **70** to produce a second control signal that controls the electrostatic potential that is applied to the second grid electrode **112**. The grid electrodes therefore receive a differential potential controlled by the beam wobble input signal ·**70** which controls the beam wobble, and a superimposed sum potential controlled by the grid modulation signal **64**.

       The frequency response of analog dose modulation performed by modulating the sum grid potential is limited by a frequency response of the grid power supply **66**. Existing grid power supplies used for electrostatic beam wobble alternately position the focal spot between two tightly controlled locations using a differential potential applied to the grid electrodes **110, 112**. The beam is stepped between the focal spots in response to a square voltage pulse beam wobble input signal **70**, at around 2-4 kHz for a 120 rpm gantry rotation and 2000 projection views per rotation.

       The harmonic content of the square-wave beam wobble input signal **70** leads to high bandwidth requirements for the grid power supply **66**, which is usually a switching power supply. Such power supplies today typically exhibit a limited analog frequency response (as opposed to a power amplifier, which would have a much higher bandwidth but would also be more costly) such that modulation of the grid potentials alone does not achieve dose modulation at the desired frequency response

of about 4Hz to 6 Hz or higher with a dose modulation dynamic range of about 8:1 to 10:1.

To achieve the desired dose modulation frequency response and dynamic range, the dose modulation controller **52**

5   cooperatively controls the grid modulation signal **64** and the filament modulation signal **60** to cascade the grid electrode dose modulation and the filament current dose modulation. For a 4-6 Hz modulation frequency, a product of the cascaded grid and filament dose modulations provides the desired 8:1 to 10:1

10   dynamic modulation range. Moreover, dynamic range can be traded off for higher dose modulation frequency.

A further benefit of combined grid and filament dose modulation is improved focus of the electron beam **94** throughout the dynamic modulation range. The x-ray tube **20** is designed to

15   operate with a small spot size at relatively high output power. Combined grid and filament dose modulation reduces the output power by cooperatively reducing the filament current and increasing the sum potential on the grid electrodes **110**, **112**. Beam narrowing due to higher sum grid potentials, results in

20   improved electron beam spot size on the anode **92** during dose modulation and consequently improved spatial definition of the x-ray beam **96** throughout the dynamic modulation range.

In FIGURE 2, the first and second summers **120**, **122** are integrated into the grid power supply **66**. However, it is

25   also contemplated to arrange these elements outside of the grid power supply, and supply the summer outputs as inputs to the grid power supply. This alternate arrangement facilitates retrofitting a computed tomography imaging system which include a grid power supply for beam wobble with dose modulation

30   capability that uses cascaded filament current and grid sum potential modulations.

Moreover, it will be recognized that the cascaded dose modulation can be performed with or without concurrent

focus switching. That is, the beam wobble input signal **70** can be turned off or omitted entirely from the computed tomography imaging system while retaining the cascaded dose modulation aspect.

5   With reference to FIGURE 1, the above-described beam modulation· can be employed in a variety of ways. In one suitable dose modulated imaging process, the dose modulation controller **52** integrates the dose delivery rate received from the dose delivery rate processor **50** over a 360° revolution of

10   the gantry **16** corresponding to an axial slice or one turn of a helical trajectory. Based on the integrated dose delivered to the subject over 360°, the dose modulation controller **52** adjusts the sum potential on the grid electrodes **110, 112** and the current through the filament **80** for the next 360°

15   revolution to maintain a generally constant radiation dose delivered for each slice or helical revolution. As the imaging progresses axially between regions of the subject having differing absorption characteristics, the x-ray beam is modulated to maintain a generally constant x-ray absorption

20   rate in the subject.

          In another suitable dose modulated imaging process, the dose modulation controller **52** continuously adjusts the sum potential on the grid electrodes **110, 112** and the current through the filament **80** to maintain a generally constant dose

25   delivery rate as reported by the dose delivery rate processor **50**. This process can adjust the radiation delivery rate during acquisition of projection data for a single slice, and is particularly suitable for large-pitch helical scanning in which a substantial axial distance is sampled during each 360° gantry

30   rotation.

          The x-ray tube **20** uses electrostatic beam deflection to implement beam wobble. Hence, the grid electrodes **110, 112** are used for the electrostatic beam wobble and are additionally

used for electrostatically dose modulating through the applied sum potential. However, in some existing x-ray tubes an electrostatic grid is not used. Rather, the electrostatic grid is replaced by an electromagnetic beam deflector.

5   With reference to FIGURE 3, an x-ray tube **20'** uses separate electromagnetic beam deflection for wobble and electrostatic electron beam constriction or aperturing for dose modulation. FIGURE 3 schematically shows components of the x-ray tube **20'** in cross-section along with associated

10 electronic components. Components in FIGURE 3 that generally correspond with components of FIGURE 2 are indicated by corresponding primed reference numbers.

        Specifically, the x-ray tube **20'** includes a filament **80'**, a cathode **84'** with a cathode cup **82'**, and an anode **92'**

15 with a rotational axis **100'** and an angled circumferential x-ray generating surface **102'**. These components generally correspond to the filament **80**, cathode **84**, cathode cup **82**, anode **92**, rotational axis **100**, and surface **102** of the x-ray tube **20**. Moreover, x-ray tube control hardware including a filament

20 power supply **62'** with an isolation transformer **86'** and a tube power supply **72'** operate similarly to corresponding elements **62**, **86**, **72** of FIGURE 2 in applying bias potentials relative to a common **90'** and generating an electron beam **94'** and an x-ray beam **96'**. A beam wobble input signal **70'** and a filament current

25 modulation control signal **60'** correspond to the analogous input signals **70**, **60** of FIGURES 1 and 2.

        The x-ray tube **20'** employs electromagnetic beam deflection rather than the electrostatic beam deflection employed by the x-ray tube **20**. Specifically, the beam wobble

30 input signal **70'** is input into a deflector power supply **120**. Inverting and non-inverting paths **122**, **124** generate power for driving electromagnets **126**, **128** based on the input signal **70'**

to steer the electron beam **94'** between two wobble position focal spots. Two electromagnets **126, 128** are shown. However, a single solenoidal electromagnet can also be employed.

Unlike the electrostatic grid electrodes **110, 112** of the x-ray tube **20**, the electromagnets **126, 128** of the x-ray tube **20'** cannot constrict the electron beam to effect a dose modulation. Rather, separate electrostatic beam constriction elements are included in the tube **20'**.

A Wehnelt cylinder **140** is driven by a Wehnelt power supply **142**. The grid modulation input signal **64** of FIGURES 1 and 2 is replaced by an analogous electrostatic modulation signal **64'** which is input to the Wehnelt power supply **142**. A negative potential applied to the Wehnelt cylinder **140** effects a reduction in an intensity of the electron beam **94'** at the anode **92'** by partial electrostatic pinchoff of the electron beam **94'**. For a sufficiently large negative potential applied to the Wehnelt cylinder **140** the electron beam **94'** can be pinched off entirely. The Wehnelt cylinder **140** is biased negatively with respect to the cathode **84'**, typically in a range of zero volts to about 3 kV more negative than the cathode **84'**.

The Wehnelt cylinder **140** performs electrostatic beam modulation based on the electrostatic modulation signal **64'** in substantially the same manner that the grid electrodes **110, 112** of the x-ray tube **20** perform electrostatic beam modulation based on the sum electrostatic potential **64**. Electrostatic dose modulation via the Wehnelt cylinder **140** is suitably combined with modulation of current in the filament **80'** to provide cascaded electrostatic and filament current controlled dose modulation. Benefits of this cascaded dose modulation described with reference to the x-ray tube **20** apply also to the x-ray tube **20'**, such as improved dynamic range of the radiation

modulation and improved spatial definition of the x-ray beam over the dynamic modulation range. Moreover, the x-ray tube **20'** optionally includes additional elements such as fluid passages for active cooling, one or more additional filaments for

5   redundancy· or for achieving different electron beam characteristics, and the like (elements not shown).

The invention has been described with reference to the preferred embodiments. Obviously, modifications and alterations will occur to others upon reading and

10   understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.